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MEMORANDUM

HUMAN TOLERANCE TO RAPIDLY APPLIED ACCELERATIONS:

A SUMMARY OF THE LITERATURE

By A. Martin Eiband

Lewis Research Center
Cleveland, Ohio

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SUMMARY

The literature is surveyed to determine human tolerance to rapidly applied accelerations. Pertinent human and animal experiments applicable to space flight and to crash impact forces are analyzed and discussed. These data are compared and presented on the basis of a trapezoidal pulse. The effects of body restraint and of acceleration direction, onset rate, and plateau duration on the maximum tolerable and survivable rapidly applied accelerations are shown.

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Results of the survey indicate that adequate torso and extremity restraint is the primary variable in tolerance to rapidly applied accelerations. The harness, or restraint system, must be arranged to transmit the major portion of the accelerating force directly to the pelvic structure and not via the vertebral column. When the conditions of adequate restraint have been met, then the other variables, direction, magnitude, and onset rate of rapidly applied accelerations, govern maximum tolerance and injury limits.

The results also indicate that adequately stressed aft-faced passenger seats offer maximum complete body support with minimum objectionable harnessing. Such a seat, whether designed for 20-, 30-, or 40-G dynamic loading, would include lap strap, chest (axillary) strap, and winged-back seat to increase headward and lateral G protection, full-height integral head rest, arm rests (load-bearing) with recessed hand-holds and provisions to prevent arms from slipping either laterally or beyond the seat back, and leg support to keep the legs from being wedged under the seat.

For crew members and others whose duties require forward-facing seats, maximum complete body support requires lap, shoulder, and thigh straps, lap-belt tie-down strap, and full-height seat back with integral head support.

INTRODUCTION

Suddenly applied accelerations may be encountered by human occupants in space vehicles as well as in conventional vehicles. Abnormal

acceleration in space vehicles will occur during rocket-powered flight, from reentry into the Earth's atmosphere, and from impact with a planet's surface on landing. Similar conditions arise from the impact of a crashing airplane with the ground.

Before the hazard of these accelerations can be appraised, the human tolerance to sudden accelerations must be known. Considerable information concerning the human tolerance to acceleration is available in the aeromedical literature. Much of this information, however, is not readily usable because the data are not reported in directly comparable forms. Other investigators also have pointed out this problem (ref. 1). In order to make this information useful, the tolerance data were collected, studied, and placed on a comparable basis at the NASA Lewis Research Center. The results of this work are described herein.

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From the literature, it is readily apparent that the human tolerance to sudden acceleration depends upon (1) the direction in which the accelerating force is applied to the body, (2) the magnitude of the accelerating force, (3) how long the accelerating force is applied, (4) how rapidly the accelerating force is applied, and (5) how the occupant's body is supported during the acceleration. The direction in which the crash force is applied to the occupant's body depends upon the seating arrangement and is an independent variable. For this reason, the results of this study are divided into four sections corresponding to the directions in which an accelerating force can be applied to the occupant's body. These directions are defined in the terminology section that follows. The remaining variables differ with the direction of the accelerating force and are therefore discussed in each of the four main sections.

Most of the pertinent live experiments reported in the literature were surveyed during this study. These references comprise a fairly complete bibliography of the reports currently available. For the convenience of others studying human tolerance to rapidly applied accelerations, a categorized bibliography of all references that were surveyed is included as an appendix to this report.

TERMINOLOGY

Direction of Forces Imposed on Occupant

The aeromedical literature shows generally that accelerations of larger magnitude can be sustained when the accelerating force is imposed perpendicular rather than parallel to the long axis of the spine. Because of this fact, the recent aeromedical trend has been to define the accelerating force in the directions parallel and perpendicular to the occupant's spine (refs. 2 and 3). In keeping with this trend, the terms "spineward" and "sternumward" will be used to define forces applied

perpendicular to the spine but in opposite directions. "Headward" and "tailward" are used to define forces applied parallel to the spine also in opposite directions. These terms are defined in the following paragraphs.

Spineward-sternumward. - In aeromedical terms, forces or accelerations acting upon the external surface of the body in the sagittal plane (fig. 1) perpendicular to the long axis of the spine are called "transverse." They are then further defined as acting "chest to back" or "back to chest" (table I). By substituting the words "spineward" and "sternumward," as shown by the arrows in figure 1, this lengthy terminology can be eliminated. Thus, in impact acceleration, forces applied to the front surfaces of the body to resist further motion are acting toward the spine, or "spineward." Conversely, accelerative forces acting on the back surfaces are acting toward the chest. A force acting in this direction is defined as "sternumward."

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Headward-tailward. - Forces tending to move the seated occupant parallel to the long axis of the spine also are shown by arrows in figure 1. Forces tending to move the occupant upward (when in the normal seated posture) have been defined as "headward" as illustrated in this figure. The term "headward" has been used previously in aeromedical literature. Table I shows that such an externally applied force has been aeromedically defined as "positive" or +G. Acceleration along the long axis of the spine in the opposite direction is conversely named "tailward" (fig. 1). "Tailward" has been substituted for the generally used "footward" so that a common term will apply to both humans and animals. This condition has been generally referred to aeromedically as negative acceleration (table I).

The use of this terminology makes possible a direct comparison between tolerance levels of seated subjects and those in the prone and supine positions without confusing terminology. In a like manner, studies of animal exposures may be compared with human exposures. If the differences in anatomical structure of various animal subjects are considered, such comparisons are physiologically justified, because studies of the effects of impact acceleration show that relative position of the extremities is not a primary factor in tolerance, if adequate support of the torso and extremities is provided. Ideal adequate support requires that the normal spinal curvatures (as illustrated in fig. 1) be maintained by holding the long axis of the spine in the erect posture. With such support, tolerance levels are determined by the forces acting directly on the torso.

Trapezoidal Pulse for Comparison of Data

In this report, a trapezoidal pulse of seat or platform acceleration (fig. 2 inset) is used as the basis for the comparison of all data.

This procedure is necessary because most of the data in the literature are presented in this way and no other basis of comparison is possible. Previous use of the trapezoidal-pulse analysis in the operation of the rocket-driven sled is described in reference 4.

The use of a trapezoidal pulse for vehicle acceleration is possible since, generally, the onset rate, total pulse duration, and peak or plateau magnitude are reported in the referenced studies. In some cases the onset rate and duration of plateau acceleration were measured from the published time-acceleration curves. The time interval during which the acceleration reaches plateau magnitude (onset time), shown as $(t_1 - t_0)$ in the inset of figure 2, and the time interval during which the acceleration returns to zero $(t_3 - t_2)$ are not considered as part of the "duration" of "uniform" acceleration $(t_2 - t_1)$.

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Classification and Description of Injuries

It was also necessary to use readily identified specific symptoms as a measure of the severity of the injury, in order to determine whether the human tolerance had been exceeded and to define the various levels of injury. The following definitions are in general use in this phase of aeromedical work.

Medically, a tolerable acceleration may be defined as one in which the subject is not debilitated or traumatically injured. Debilitation is a state of abnormal weakness, languor or feebleness (refs. 5 and 6). This effect does not necessarily result from wounds or lesions. Traumatic injury as defined for this report includes wounds and lesions but does not include superficial cuts and wounds, bruises, or strap abrasions, as such injuries would not deter a rational escape attempt. Either debilitation or traumatic injury then defines an exposure that exceeds the limits of voluntary tolerance.

In further classifying degrees of injury, a modification of the scale developed by the Cornell Crash Injury Research Group has been used. This scale is reproduced in table II, and the degrees of injury defined as undebilitated and uninjured, moderate, and severe injury are indicated in the right column. Moderate injury includes slight injury of extremities, short-time unconsciousness, dislocation, and simple spine fractures. Severe injury includes dangerous-to-life injuries such as dangerous hemorrhages, spine injury, abdominal and thoracic injury, multiple fractures, concussion, and long-time unconsciousness.

For obvious reasons, animal experiments have been used to define the severe-injury threshold, the moderate- and severe-injury boundary. These limits are intended to indicate the margin between voluntary human tolerance limits and possible severe but survivable injury. Caution must be

exercised in using this boundary interchangeably between animal and human subjects because of the admitted anatomical and physiological differences between the human and animal bodies. As shown later, however, survival of falls and the results of a few human experiments do indicate that the severe-injury threshold may be reasonably applicable to human survival limits.

HUMAN SURVIVAL LIMITS INDICATED BY PUBLISHED LITERATURE

There are certain limitations to the tolerance data presented in the literature that must be taken into account when using this information to appraise the hazard of any abnormal acceleration. For a rigorously correct appraisal of the hazard, the acceleration pulse imposed upon parts of a person's body such as head, chest, and hips should be used for defining the tolerance and injury levels and for appraising the hazard of crash accelerations. Unfortunately, in many instances, the investigators studying human tolerance to acceleration were not able to measure accelerations at these locations. Instead, the tolerance data are presented on the basis of the acceleration applied to the vehicle or platform to which the seat was attached. In most cases the amplification or attenuation (dynamic response) that results from the reaction of the seat to the floor acceleration was eliminated by making the seat structures overly strong and rigid so that the acceleration of the seat was the same as that of the floor. With this approach, however, the tolerance data can be used only for circumstances in which the seats and harness are similar to those used for the tolerance studies. This method of presenting the data limits its usefulness, because the potential injury resulting from a given impact acceleration applied to the floor of the vehicle cannot be appraised unless the dynamic response of the seats in that particular vehicle is known and unless the restraining harness is dynamically similar to the harness in the tolerance studies referred to in this discussion.

The limits presented in the following paragraphs apply to seated occupants only when held by the restraining harness described for each of the four directions of the applied accelerating force. If the occupant is placed in other than a seated posture, the effect of the necessary changes in the harness arrangement and the possibility of a change in tolerance must be carefully evaluated before the limits presented can be considered to be applicable.

These limitations could be eliminated by determining the individual response curves for seat, harness, and occupant. With response curves for various seats and harness arrangements, it would then be possible to determine the net acceleration imposed on the head, chest, or hips of an occupant for any combination of seat, harness, and floor acceleration.

I - Survival of Spineward Accelerations

Spineward accelerations will be discussed first because they are of the greatest interest. In the forward-facing position, which is the most common today in the United States, abnormal accelerations usually impose spineward rather than sternumward accelerations.

Indicated limits with maximum body support. - The combinations of acceleration and the time duration of the acceleration that human and animal subjects have found tolerable or have survived with either moderate or severe injury are shown by the three zones in figure 2. The solid line marks the boundary between exposures that human beings have voluntarily tolerated without being weakened or injured. The dashed line is the boundary between exposures that produce moderate or severe injury. Exposures that fatally injure human subjects have not been measured and cannot be defined on this chart. A large number of spineward experiments have been reported in the literature. For the purpose of this study, only those exposures that could be used to define the boundaries of the various injury levels have been plotted.

The voluntary-human-tolerance boundary shows that subjects have endured maximum uniform accelerations of 45 G's for 0.044 second with no injurious or debilitating (weakening) effects (fig. 2). Since the subject was not injured or weakened by this impulse, healthy persons should be able to survive comparable exposures, for example in a crash, and to make a rational escape attempt immediately thereafter. The acceleration onset rate in all of the tolerable exposures was about 500 G's per second.

When the duration of the plateau (or uniform) acceleration exceeds 0.044 second, the limiting magnitude of voluntary exposures decreases rapidly. As further shown in figure 2, the upper limit of voluntary tolerance decreases to a magnitude of 40 G's for intervals of 0.1-second duration and to 25 G's when the interval is increased to 0.2 second.

As the duration of the uniform acceleration is increased still further, the upper limit of tolerance continues to decrease. When the subject was restrained by chest, arm, leg, and helmet straps in addition to the experimental lap, shoulder, and thigh straps, exposures of 13 G's for 0.6 second and 10 G's for 1.6 seconds could be tolerated (ref. 7). In the experiments just cited, all straps were drawn up tighter than a person usually would wear them. Reference 8 states that the straps were pulled to a static tension of 25 to 50 pounds. Strap slack thus was virtually eliminated. Complete details of the harness and of the 3-inch webbing used in the human experiments are shown in figure 3 (group H). Figure 4 also indicates the amount of extremity restraint required in the human experiments. Details of the thigh strap attachment to the lap strap are shown in figure 5.

The limits for moderate-injury exposures are shown by the dashed line that forms the boundary between the moderate- and severe-injury zones in figure 2. The upper limit of moderate injury was established with a hog subject that endured 160 G's for about 0.004 second. These values were measured on the seat bottom from the data of figure 6 (reproduced from ref. 9). Total stopping force (total restraining-strap tensions) recorded in this experiment was 10,700 pounds, or 116 times the subject's normal weight. The total strap tensions thus did not equal the forces expected from the acceleration value. No explanation was given for this difference.

In exposures of longer duration, chimpanzee subjects have endured plateau accelerations of 56 G's for 0.037 second and 43 G's for 0.110 second. Reference 9 reported that the injury to these hog and chimpanzee subjects ranged from moderate or none to severe shock and concussion, depending upon the severity of the exposure. The subjects appeared normal within two or three days following the exposure. Subsequent autopsy of both hogs and chimpanzees revealed no injuries that would require an extensive growth repair process.

The single human exposure to 25 G's for 0.93 second (fig. 2) completes the limit of the moderate-injury area. A velocity change of 750 feet per second resulted from the uniform-acceleration portion of the acceleration pulse. The total 857-foot-per-second velocity change was experienced by the subject in 1.1 seconds (ref. 7). The harness used for this experiment was similar to those used for the tolerable and other moderate-injury experiments plus the chest strap, helmet straps, and multiple arm and leg lashings described in reference 7. In this experiment, the subject was debilitated but conscious. Although the subject could stand erect momentarily, and could control hand and arm movements following release of the straps, he could neither see nor maintain a standing posture. The subject returned to normal duty in five days.

The human exposure to accelerations of 200 G's shown above the boundary of moderate and severe injury is an estimated value obtained from a fall from a building. This person survived the fall, and injuries were essentially moderate. The magnitude of the moderate- and severe-injury boundary may thus be greater than indicated for these durations. The boundary was not shown at this magnitude, however, because the value is an estimate, not an experimental value.

Exposures to accelerations of the magnitudes and durations defined by the severe-injury zone are hazardous to life. Recovery may require surgery and a long-time growth process to repair the damage. Sufficient experimental data have not been obtained to indicate the magnitude of the lethal boundary of severe injury.

Variation of limits with vehicle onset rate. - The fact that onset rate, the rate with which the accelerating force is applied, affects the human tolerance and injury levels is discussed in references 9 to 11. Figure 7 is arranged to show the effects of the onset rate, as measured on the vehicle or on the seat pan, on the magnitude of survivable acceleration. The onset-rate portion of the acceleration history is illustrated by the shaded triangular portion of the trapezoidal-pulse inset in figure 7 from time t_0 to t_1.

In human exposures, an acceleration of 45.4 G's was reached with an onset rate of about 500 G's per second (fig. 7). This exposure produced no debilitating or intolerable effects on the subject. Only a rise in blood pressure and pulse rate typical of moderate exercise was noted. An exposure to 35 G's at an onset rate of 600 G's per second was also tolerable. When the onset rate was increased to 1370 G's per second, definite signs of shock were produced at plateau levels of 38.6 G's. One of the subjects fainted immediately after the run, indicating that the exposure was slightly above the uninjured-undebilitated limits. The other subject was undebilitated.

In studying the effect of onset rate on tolerance to maximum accelerations, reference 12 states that the rate and magnitude effect appeared when the plateau acceleration exceeded 30 G's and the rate of onset 1000 G's per second. Onset-rate magnitude levels exceeding this combination produced unpleasant pressure sensations from restraining straps. Pallor, drop in blood pressure, increased pulse rate, and occasional spasms within the eyes also were observed.

Experiments with chimpanzee subjects (fig. 7) indicate that survival of onset rates exceeding the human exposures discussed in the last paragraph is possible. The onset rates for the two survivable animal exposures shown were obtained from reference 9. These data show that, with an onset rate of 1060 G's per second, a plateau acceleration of 28.2 G's was sustained with military lap- and shoulder-strap restraint. When the onset rate was increased to 3400 G's per second, the tolerance to acceleration was 35 G's with the subjects restrained by lap, shoulder, and thigh straps. Although these high rates of onset produced cardiovascular shock and minor harness abrasions, the subjects had resumed their normal activities within several hours following the experiment. These animal experiments, therefore, indicate that humans, adequately restrained, may receive only moderate injury when exposed to onset rates of 3400 G's per second and maximum magnitude of 35 G's. However, they would not be capable of making a rational escape attempt immediately after such exposures.

The possibility of human survival of onset rates above 3400 G's per second and maximum magnitudes of 35 G's is indicated in experiments with hog subjects. Rates up to 13,000 G's per second (160 G's peak magnitude)

have been reported in reference 9, wherein the subject was described as able to stand up 10 minutes after the experiment. Immediate X-ray examination of the subject revealed no indication of fractures. Autopsy of this subject 10 days later showed no injury from the rapidly applied force. Higher rates of onset are quoted for other exposures in this series. The published data, however, do not provide sufficient details of the method by which the onset rates were obtained. Onset rates from the hog studies, therefore, are not included in figure 7.

<u>Variation of limits with amount of body support</u>. - The tolerance and injury limits described in the preceding sections were endured by subjects held by a restraining harness that included lap, shoulder, and thigh straps. If thigh straps are not used, the voluntary human tolerance is reduced to the extent shown by figure 8. The lower line on the figure shows that, when the restraining harness does not hold the pelvis in place, all exposures above 18 G's are intolerable. The forms of harness studied that provided only lap and upper-torso restraint varied from the lap- and shoulder-strap combination (ref. 12) to a 16-inch-wide abdominal girdle (ref. 10) and a complete abdominal vest (ref. 13).

The most widely used of such restraints is the military lap- and shoulder-strap combination illustrated in figure 3 (group D). Subjects equipped with this arrangement would not volunteer for exposures above 11.3 G's and 0.28-second calculated duration, the smallest value on the lower line of figure 8. Reference 12 states that exposures up to 17-G peak magnitude were endured with this restraint. The published data, however, do not show accelerations of this magnitude. The subjects reported a marked impact to the shoulders and abdomen from these exposures (table III, group D). High-speed motion pictures showed that the accelerative force raised the seat belt from the pelvis to the upper abdomen and lower rib region. Application of the restraining force to this region of the body produced sharp pains in the rib regions.

Other attempts to increase tolerance to spineward accelerations by using a vest or wide belts were equally ineffective. Experiments with dummy subjects on the rocket sled (ref. 12) showed that, without thigh straps, the vest-type upper-torso harness forms an ejection chute, and the inertia of the legs and thighs pulls the pelvis and lower torso out from beneath the lap belt. This action acutely flexes the lower spine, and the belt strikes the solar plexis. Motion pictures also showed that, when a 16-inch-wide girdle around the chest and abdomen was used, the subject's torso folded forward around the girdle so that head, shoulders, and arms, as well as legs and pelvis, flexed forward (ref. 10).

Thigh straps overcome the disadvantages of the lap-strap and upper-torso restraint just discussed. Reference 12 attributes the benefits of the thigh straps to their action in preventing headward rotation of the lap strap. The lap strap is held on the pelvis, thus applying the major

portion of the accelerating force to the pelvic-girdle structure instead of the soft abdominal region.

Results of studies with combinations of standard military harness and thigh straps are shown by the upper curve of figure 8 (group L). Maximum exposure of 32 G's and 0.18 second was sustained. Thus, the addition of thigh straps increased the tolerance limits to almost three times (32/11.3 G's) that acceptable with lap and shoulder strap alone. The subjects' remarks concerning exposures with this group of restraints (groups L, K, and I of table III) are uniformly favorable.

Eliminating the sources of discomfort reported in table III enabled subjects to endure a maximum of 45 G's (fig. 2, group H). The difficulties in the experiments with the modified military harnesses listed in table III were reduced by making all straps of double-thickness, number 9 nylon webbing, 3 inches wide (see fig. 3, group H). Table III, group H, shows that the subjects' previous objections were overcome with this restraint.

A photograph of the torso harness consisting of lap, shoulder, and thigh straps is shown in figure 5. This photograph shows how the thigh straps are looped over the lap-strap buckle, then passed around the thighs and under the buttocks to the rear corners of the seat. Details of the webbing materials and dimensions used in the component straps of this particular restraint are shown by the schematic diagram, group K, in figure 3. Figure 4 shows a subject restrained by a harness similar to that shown in figure 5 plus a slack emergency chest strap. Arms, legs, and feet are also shown lashed down.

II - Survival of Sternumward Accelerations

Occupants of an aircraft may be seated facing aft as well as forward. Under these circumstances, in a landing impact the acceleration would drive the occupant against the seat back, and the accelerating force would act sternumward on the occupant. Since sternumward acceleration acts against the back surface of the body instead of acting against the front surface as in spineward acceleration, the injury limits may differ. The sternumward limits as indicated by the literature are discussed in this section.

Indicated limits with maximum body support. - Magnitude- and time-duration limits reported in the literature for sternumward accelerations sustained by human and animal subjects are summarized in figure 9. Coordinate units, injury levels, and requisites for plotting data from the literature are similar to those used for the spineward accelerations discussed in the preceding section.

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Although a restraining harness is not needed for purely sternumward impact forces, the experimental data show that the maximum tolerance can be obtained only when the seat is properly padded and the arms, legs, and head are held in place. In the majority of the experiments summarized by figure 9, the seat back was padded with 1/2 inch of felt (fig. 10) and extended to the subject's full height to provide a head rest (fig. 11). The head rest was also padded with a pig-hair cushion between the seat back and the occupant's helmet (fig. 12). Legs were lashed down as shown by figure 13 to prevent dislocation of joints and to keep the subject's legs from striking the seat structure. This figure also shows the subject gripping vertical hand-holds.

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Supported as just described, human subjects have voluntarily and without injury endured uniform accelerations of 35-G magnitude for periods up to 0.1 second (fig. 9) on the rocket-driven sled. The total duration of the exposure shown in figure 9 was 0.16 second. This exposure represents a complete stop from 72 miles per hour in a distance of 8.2 feet (ref. 12). For safety reasons, the restraining harness in this run was a lap strap and shoulder harness. A similar run was made later with only a lap strap. A lap strap would be needed for rough air and possible spineward exposures even though occupants are seated facing aft.

Similar magnitudes of acceleration (30 G's or 4950-lb force) had been endured in previous experiments on the German giant swing (ref. 10). However, the values reported were peak magnitudes, and the total pulse was less than 0.1-second duration.

Voluntary human exposures of longer duration on the rocket sled (ref. 14) have been endured for plateau durations of 30 G's for 0.12 and 21 G's for 0.19 second. As shown in figure 9, tolerable plateau magnitudes decreased as the duration increased.

These limits apply when the onset rate is 1150 G's per second or less. Onset rates below 500 G's per second were found to be preferable. The effect of onset rate is discussed fully in the next section.

Voluntary human exposures sternumward were not carried to the limits of magnitude and duration studied in spineward exposures. It has been concluded, however (refs. 12 and 9), that voluntary tolerance to sternumward acceleration should equal the tolerance to spineward acceleration. At comparable exposures, no difference in subjects' reactions to spineward or sternumward accelerations was found.

Animal exposures are used to define the boundary of moderate and severe injury. Figure 9 shows that chimpanzee subjects have survived exposures of about 49 and 34 G's for respective periods of 0.064 and 0.13 second. These exposures produced definite shock, although the subjects resumed normal activity the following day. Sternumward animal exposures

of duration longer than those just quoted have not been reported in the literature. However, since sternumward chimpanzee exposures produced symptoms similar to those from comparable spineward acceleration, it would appear that the sternumward moderate-injury limit could be extrapolated along the moderate- and severe-injury boundary of spineward exposures. Accordingly, the data are extrapolated along the wide band to 25 G's and 0.96-second duration. The extrapolation is taken from the moderate-injury level of figure 2.

Exposure of hog subjects to rapidly applied acceleration of high magnitudes indicates that the human body may survive, with severe injury, magnitudes larger than the chimpanzee exposures, when the durations are very short. Figure 9 shows that hogs have survived exposure to 140 G's for about 0.002 second. Reference 9 reported injury from this exposure as subpleural hemorrhage, visceral and parietal.

Human survival of accelerative forces in excess of those recorded in the hog exposures is indicated by the data points at 160 G's in figure 9. These data represent survival of human beings falling from heights of 50 to 150 feet (ref. 15). The calculated time and acceleration magnitudes shown indicate that impact forces considerably in excess of the experimental values obtained can be survived by human and animal subjects when the body is adequately supported. The local stresses resulting from accelerations of these magnitudes and durations are so great, however, that the random uncontrolled circumstances of the incident that create large local stresses may determine whether the subject survives. Several examples are given in reference 15 of exposures that were essentially similar with respect to acceleration and body support but in which minor random variations determined whether the exposure was survivable or fatal.

Variation of limits with vehicle onset rate. - Figure 14 shows the effect of sternumward onset rate (or time rise to maximum acceleration) on injury levels of human and animal subjects. The onset-rate portion of the acceleration history is illustrated by the shaded triangular portion of the trapezoidal pulse extending from time t_0 to t_1.

Again, it should be noted that the plotted exposures were sustained only with maximum body support as described in the original experiments. A maximum human exposure of 35 G's at an onset rate of 1156 G's per second is shown in figure 14. This exposure was reported as undebilitating (ref. 12).

The second human exposure plotted in figure 14 was accomplished at an onset rate of 530 G's per second, reaching a plateau value of 31.5 G's. This exposure was undebilitating also. Reports of the subject's condition following each of these runs were similar: a stiff jolt to the head as from a blow in boxing, the feeling of a heavy impact, and a fleeting

headache on shaking the head. The jolt reported by the subject exposed to 35 G's at 1156 G's per second, however, apparently was more severe than reported from the 31.5-G exposure at 530-G-per-second onset rate.

Chimpanzee exposures plotted in figure 14 indicate that human beings may be able to tolerate onset rates greater than those just described. Debilitating effects of shock may result, however. Chimpanzees have survived maximum exposures of 34.2 G's at 3350 G's per second and 47.4 G's at 1065 G's per second. Although otherwise uninjured, chimpanzees exposed to 34.2 G's at 3350 G's per second sustained cardiovascular shock (ref. 9).

These data then indicate that onset rates up to 1100 G's per second can be tolerated up to 35 G's without weakening the subject. Onset rates of 3000 G's per second to the same magnitudes, however, probably will produce symptoms of shock.

<u>Body support used for sternumward experiments</u>. - The restraining and protective equipment necessary for sternumward impact forces is relatively simple, as indicated by the previous description. Apparently because of this simplicity, no data are reported in the literature showing the effect of variations in the amount of body support on tolerance. Consequently, the effect of variations in restraint cannot be discussed. The support used is described in greater detail in the following paragraphs for readers that need detailed information. Figures 10, 12, 13, 15, and 16 illustrate the various items found necessary in the experiments with human and animal subjects.

In all of these experiments, the seat and its attachments were sufficiently rigid to resist deformation and failure. Figure 11 shows the experimental seat configuration and structure reproduced from reference 12. References 14 and 16 reported that this seat was designed to withstand 150 G's at the breaking limit and weighed 56 pounds.

Figure 12 illustrates the full-height seat back and the integral head rest necessary for head support when seated facing to the rear. Details of head restraint and height of seat back with respect to subject's head also are shown in this figure. The subject is wearing a crash helmet with a pad of pig's hair between seat and helmet.

Relative movement between head and body was satisfactorily controlled with the head-rest cushion shown in figure 12 and a 1/2-inch-thick felt pad on the seat back. The back padding, such as shown in figure 10, was adequate for the maximum-exposure runs plotted in figure 9.

For basic harness, the sternumward exposures consisted of lap and chest straps of 3-inch webbing. A subject so restrained for an experiment is shown in figure 13. This figure also illustrates the extremity

restraints used in these experiments. It was necessary to strap feet to the canted foot rest in all runs.

Vertical hand grips provided sufficient arm restraint in all except the maximum-exposure runs. In these runs the combination of the subject's hand grip and the wind force was not enough to keep the impact from swinging the subject's arms around behind his back. Figure 16 shows the seat structure and the method of attaching helmet, chest, and lap straps to the seat for the aft-facing experiments of reference 14.

III - Survival of Headward Accelerations

In addition to spineward and sternumward impact forces, the occupants of a suddenly accelerating aircraft may also be exposed to impact forces acting headward. For example, such impact forces occur when an airplane strikes a surface inclined to its path, when an airplane has a large sinking speed relative to the runway or when it strikes an upslope during takeoff. In such a crash, the direction of the accelerative forces with the respect to a seated occupant will be such that his spine will be compressed and subjected to column stresses. Crash-injury studies show that such stresses frequently cause spinal injuries. The maximum headward accelerations that human beings can tolerate are discussed in the following paragraphs from data reported in upward ejection-seat experiments.

<u>Indicated limits when supported by lap and shoulder straps.</u> - The maximum magnitude- and time-duration limits reported in the literature are summarized in figure 17 based on the duration of the uniform acceleration for a trapezoidal pulse as shown by the inset. In most cases to date, body support has consisted of the conventional lap and shoulder straps plus face curtain or partial arm rests, and generally the human subject has been seated on a parachute seat pack. Additions to the present harness such as chest and thigh straps have been proposed; however, these possible improvements have not been tested.

With a face curtain, the weight of the arms is carried partially by the curtain. The curtain also helps to hold the head and neck in a normal seated posture. The essential details of this system are shown by figure 18. Not enough data were available, however, to make a complete comparison on the basis of a single type of restraint. Consequently, it has been necessary to compare data using more than one mode of restraint. Injury levels are plotted consistently with injury levels previously defined in table II.

Figure 17 shows that human volunteers, with lap strap, shoulder straps, and face curtain have endured uniform seat accelerations of 16 G's for periods up to 0.04 second in catapult seat experiments. The

record from which this value was taken is reproduced as figure 19 (from ref. 17). This figure shows that the subject endured a total velocity change of 55.8 feet per second in a distance of 40 inches. Similar magnitude-duration values have been plotted from other catapult seat experiments reported in reference 17 to complete the area of uninjured, undebilitated human exposures shown by figure 17.

Since the 20-G median value of the shaded band in figure 17 has been assumed the safe design limit for ejection-seat performance since World War II, human exposures have exceeded this value infrequently. Data used to establish this line were obtained from a study of the compressive strength of the human spine. In this study (refs. 10 and 11) fresh vertebrae from cadavers were installed in a compression machine and loaded just to the fracture point. The conclusions from these experiments were that a 20-G acceleration of 0.005- to 0.5-second duration would be tolerable. As shown by the band, higher magnitudes can be tolerated if the duration is less than 0.005 second. Although other reports indicate that peak accelerations of 25 to 33 G's have been sustained (refs. 18 and 19), the duration of these accelerations is not listed. Hence, these values are not included in figure 17.

Data from animal exposures plotted in figure 17 indicate that forces considerably in excess of the uninjured human tolerances may be survived with moderate injury. Hog subjects that had been exposed to plateau accelerations of 110 G's for 0.002 second without permanent injury were completely normal in a few days (ref. 9). The base duration of these 110-G pulses, however, was about 0.030 second. Thus, accelerations below 110 G's could be survived for intervals greater than 0.002 second. Accordingly, this exposure indicates that a 42-G pulse could be tolerated for 0.007 second.

In these hog experiments, the subject was seated on a 2-inch Styrofoam cushion. The harnesses were lap and shoulder straps; but, since the subject's torso was in the supine position, a small amount of back support was provided that would not be present in the upright seated posture. The total velocity change was about 21.1 feet per second.

In experiments of longer duration, chimpanzee subjects have endured exposures of 42 G's for 0.048 second and 28 G's for 0.14 second without permanent injury (ref. 9). These exposures were conducted on a rocket-propelled sled with the subject supine and restrained by a parachute-type harness.

<u>Variation of limits with vehicle onset rate</u>. - Variation in the onset rate can affect the magnitude of tolerable headward acceleration also The effect of changes in the onset rate on tolerance is discussed in the following paragraphs.

Figure 20 shows the acceleration-magnitude and onset-rate tolerances similar to the onset-rate spectrum plotted for sternumward accelerations. Cushioning and any restraint in addition to standard U.S. military lap and shoulder straps are indicated by the symbols in this figure. The onset rates for all human experiments were measured from acceleration curves published in the respective references. Since data curves from the animal experiments have not been published, onset rates for the chimpanzee exposures were calculated from tabular data.

The maximum acceleration of 17 G's was reached at an initial onset rate of 180 G's per second. Acceleration at this rate of increase was maintained up to a plateau of about 10 G's (figs. 19 and 20). The multiple-charge catapult then accelerated the seat and occupant to about 17 G's. This multiple-step acceleration resulted in an easily tolerable total velocity change of 55.8 feet per second in a distance of 40 inches (ref. 17). E-345

In experiments where a single charge was used, the initial acceleration increased at a rate of 500 G's per second (fig. 20). Peak acceleration at this rate was 9 G's, and a velocity change of 42.5 feet per second in a distance of 60 inches was recorded. The conclusion of reference 17 indicates that an acceleration exceeding 10 to 11 G's at 500 G's per second jolted the occupant severely. In other catapult seat developments (M-3 catapult, ref. 20), plateau accelerations of 17 G's have been tolerable when reached at a constant onset rate of 115 G's per second.

Maximum onset rate of headward acceleration published in the literature is shown as about 1300 G's per second to a peak magnitude of about 12 G's. Reference 21 reported this as an extremely severe jolt to the occupant. Apparently this was much more uncomfortable than the multiple acceleration increase (tapered charge) to 17 G's reported by reference 17.

Data from animal experiments are used to indicate the effects of greater onset rates. These data indicate that magnitudes of 50 G's may be survivable with rates up to about 1000 G's per second (fig. 20). Reference 9 reports that chimpanzee subjects accelerated at this rate survived with no evidence of permanent injury. These subjects, however, were very carefully strapped to the rocket sled. The literature indicated that extreme care must be exercised in positioning the subject to sustain such forces.

Data tabulated from exposures of hog subjects indicate that onset rates even greater than those recorded in the chimpanzee exposures may be survivable. Reference 9 reports survival of 79 to 86 G's at onset rates of about 33,000 G's per second. Although these exposures indicate that severe injury may be incurred, they also indicate that human beings may survive onset rates greater than the limits indicated by the uninjured, undebilitated human exposures.

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Variation of limits with amount of body support. - The way in which the occupant's body is held is important in enduring headward impact forces also. The chain of circumstances that leads to injury when a person is subjected to headward impacts has been discussed in the literature. A brief discussion of such injuries is included in reference 22.

Not enough data are available to plot a figure indicating the effect that various modes of body support have on the headward accelerations that can be tolerated. The following discussion, however, gives an indication of the benefits that can be obtained by properly restraining a person's body during headward accelerations.

In the initial German investigations, the importance of adequate restraint was not realized, and the subject was restrained with only a lap strap. In this investigation three of four subjects were injured. Reference 23 showed that the injury was not noticed until fourteen days after the exposure.

Early British experiments produced the same results (refs. 24 and 25). One subject was injured at about $3\frac{1}{2}$ to 4 mean G's. The injury was discovered three or four days later. The appearance of the wedge-shaped fracture that resulted from the early British experiment is shown in figure 21 from reference 25. The severity of this injury apparently was comparable to that of the injuries reported in the German experiments.

Experiments in which the arms and shoulders were supported by arm rests in addition to lap and shoulder straps increased the accelerations that could be tolerated (ref. 26). The arm rests were designed to hold the occupant's arms in place (fig. 22). In this case the tolerable peak accelerations varied from 10 to 12 G's. Supporting the arms and shoulders with arm rests reduced the weight supported by the lower spine and thus increased the tolerance. The accelerations quoted here represent faired plateau values, and the original data show isolated transient peak values above the values quoted. The subjects' complaints during these runs and motion pictures both indicated that severe flexion of the neck was the limiting ractor. The extent of this flexion is shown by figures 23(d) to (f) and figure 24. Severe stress on the lower (lumbar) vertebrae apparently was not a problem in these runs.

The values of tolerable acceleration quoted in the previous paragraph for faired-plateau acceleration values are about half those quoted in reference 11 for short-duration exposures using lap and shoulder straps but no arm rests. These German experiments (data plotted in fig. 25) indicated that "back" and "abdominal" straps arranged to keep the vertebrae in proper alinement increased the acceleration tolerance to a 23-G peak for accelerations of very short duration. Current lap and shoulder straps are the modern counterpart of the "back" and "abdominal" straps used by

the Germans in these experiments. Figure 26 shows a subject in position with the German straps. Another German investigator (ref. 23) also reported that peak accelerations of 23 G's were tolerable with similar harness.

Additional German experiments described in reference 11 also indicated that arm rests in addition to the lap and shoulder straps were desirable and that they increased the voluntary tolerance to approximately 28 G's (fig. 25). Reference 27 also reported 28-G peaks tolerable with arm rests and lap and shoulder straps. The subjects reported these exposures less severe than 23-G exposures without arm rests.

It is thought that the differences between the results of reference 26 and those in references 11, 23, and 27 are the result of the manner in which the results are stated. The German values are apparently short-duration peak values, whereas the data of reference 26 are faired-plateau values. Short-duration peaks of about 23-G and 28-G peak magnitude are shown by figure 25, although the faired-plateau values would be about 12 and 15 G's, respectively.

Providing support for a person's neck and head will reduce the severe neck flexion mentioned previously (ref. 26). A face curtain such as shown in figure 18 helps to hold the occupant's head in place. This figure shows the face curtain and occupant's posture before a run. Providing this support in addition to lap and shoulder straps but no arm rests increased the tolerance to 17.9 G's (ref. 26) as compared with 10 to 12 G's tolerance with lap and shoulder straps plus arm rests. This acceleration represented a velocity change of 56 feet per second in a distance of 40 inches, as shown by the seat-acceleration curve of figure 27.

All of the experiments just described indicate that maximum tolerance of headward acceleration can be obtained only by keeping the entire spinal column in proper alinement. With support by lap strap only, injury has occurred at only 3 to 4 G's. Minimum satisfactory body support appears to consist of the standard lap- and shoulder-strap restraint. Further relief of the body weight borne by the lumbar vertebrae through the proper use of arm rests will increase tolerance to headward accelerations. Additional support such as thigh and chest straps to maintain proper vertebral alinement of the spine may increase the tolerance to still higher values.

The exact effect of tensing one's muscles in an attempt to increase his ability to endure impact forces is not definitely known. Some people contend that tensed muscles reduce amplification of the applied acceleration, others contend relaxed muscles reduce the amplification. Generally, however, all sources indicate that sufficient muscle tension to hold the torso against the seat back is desirable.

Cushioning comparable to the military parachute seat pack was the most satisfactory of seat pads studied but may not be the optimum. Deep, soft cushions were found to amplify the acceleration imposed on the occupant (ref. 28). Maximum exposures were tolerable, therefore, only with very little cushioning between the seat pan and the occupant. The padding shown by the cushion in figure 22 appears to be satisfactory.

IV - Survival of Tailward Accelerations

Indicated limits with various types of support. - Tailward accelerations, although occurring less frequently than spineward, sternumward, or headward accelerations, do occur occasionally. The anatomical stress that these accelerations impose upon the occupant depends especially upon the restraining-harness arrangement used. If a lap strap alone is used during a tailward acceleration, the entire spine will be in tension. If shoulder straps are used in addition to the lap strap, the spine will be compressed up to the shoulders and the neck alone will be in tension. There were no experiments cited in the literature in which the entire accelerating force was applied through lap or pelvis straps alone. Consequently, it was necessary to use data obtained from experiments with varying methods of restraint. Some of the limits thus may represent headward accelerations with respect to the lower spine. The subject's neck, however, was in tension in all experiments cited, and from this standpoint the data represent only tailward accelerations. The arrangement of the harness and the manner in which it transmits tailward acceleration to the body therefore must be considered when the results of this study are used.

Prolonged tailward accelerations produce what is commonly known as "redout" as compared to blackout from headward accelerations. Redout has been interpreted as an indication of possible intracranial hemorrhage, because acceleration forces exceeding 4.2 G's have been reported to cause mental confusion, intense head pains, and disorientation (refs. 29 and 30). Experiments with human beings subjected to impact accelerations on ejection catapults, however, have shown that much greater accelerations are tolerable for small time intervals before these symptoms appear. In the 28 exploratory experiments described in reference 30, most subjects noted only the jerk of the shoulder straps when exposed to a maximum of 7.0 G's. No evidence of mental confusion or disorientation following the runs was reported. Other experiments (ref. 31) were reported in which human subjects experienced a maximum exposure of 8.5 G's with no more than a hard, disagreeable jolt. In these experiments, the standard military lap- and shoulder-strap arrangement used in reference 30 had been modified by locating the lap strap across the thighs. From these studies, it was concluded that intracranial hemorrhage was not as hazardous as had been previously supposed.

On the basis of the conclusions from these exploratory studies, an explosively propelled catapult was built that exposed subjects to a maximum seat acceleration of approximately 10 G's on a downward ejection tower (refs. 32 and 33). In a series of 35 live ejections, peak accelerations of 8 to 11 G's were reported easily tolerable with restraining harness consisting of lap belt, shoulder straps, and lap-belt-tie-down strap. The lap-belt-tie-down strap was used in place of the lap strap across the thighs, since such a lap-strap location would not be a practical means of restraint for spineward accelerations.

Data from this series of 35 human experiments have been used in plotting the voluntary human tolerance limits shown in figure 28. References 32 and 33 state that the M-4 catapult described in reference 20 was used in the 35 live experiments. Therefore, the points on the human tolerance curve between 4 and 10 G's have been measured from the M-4 acceleration-time curve. In terms of maximum human exposures, the acceleration magnitudes of part of these points may be considered conserative, because the published M-4 acceleration curve is the average of 47 development runs. Reference 32 shows that the peak catapult acceleration ranged from 7.6 to 10.5 G's for an ejected weight of 309 pounds.

Figure 28 shows that human beings have voluntarily tolerated tailward accelerations of about 10 G's for about 0.004 second. When the acceleration lasts about 0.1 second, the voluntary tolerance decreases to about 7 G's. The tolerance then decreases rapidly with increased duration.

On the basis of the results of these ejection-tower studies, the investigators concluded that the reported forces were safe for human beings and also that greater accelerations can be tolerated and a considerable margin of safety exists. These conclusions were verified by reports of seven in-flight ejections (refs. 32 and 34), in which experienced parachutists generally described the ejection force of 10 G's as very mild. The experimental values of figure 28 forming the voluntary human tolerance limits to tailward acceleration therefore appear to be definitely conservative.

Although animal tolerance to tailward acceleration may not be exactly the same as for human beings, data from such experiments may indicate how much the human tolerance limits may be increased before any form of injury would occur. The middle curve in figure 28 shows the durations of accelerations that dogs have sustained with only petechial hemorrhages (small blood spots), generally in the mucous sinus membranes. These dogs endured accelerations varying from 15 G's for 0.05 second to 7 G's when the duration was increased to 1 second. Autopsy of the subjects after various survival periods following the exposure revealed no indication of intracranial hemorrhages (refs. 35 and 36).

On the basis of these experiments, the investigators concluded first, that petechial hemorrhages occur in the frontal sinuses of dogs exposed to tailward acceleration before any other injury; and second, that a wide margin of safety exists between this end point and the dangers of intracranial hemorrhage.

In other experiments (ref. 36) dogs have survived peak tailward acceleration of 50 G's. The total pulse duration was reported to be 0.05 second in these drop-tower experiments. These data are not included in figure 28, however, because curves were not published from which plateau acceleration-time values could be measured.

Exposure of adult goats to 5 G's for 15 seconds (ref. 37) indicated an injury threshold comparable to that reported in the dog experiments. The only indication of injury reported in nine subjects was one case with symptoms of fluid in the glottis. The threshold of any detectable injury in figure 28 thus has been extended to 5 G's and 15 seconds.

The boundary between moderate and severe injury is shown by the upper curve in figure 28. Chimpanzees have survived exposure to a plateau of 60 G's for approximately 0.007 second (ref. 9). When the duration of the exposure is increased to 0.15 second, the magnitude of accelerations that can be survived without the hazard of severe injury decreases to 50 G's.

One survivable human experiment has approached the maximum acceleration exposure of animal subjects just discussed. This human exposure resulted from a spineward acceleration with complete body, arm, and leg restraint (ref. 7). Since no head restraint was provided, the subject's head rotated forward and downward into a tailward G-position as shown by the inset of figure 28. The subject noted only signs of congestion in the head from 13 G's for 0.6 second. No symptoms of intracranial hemorrhage were observed in the subject; however, the full hydraulic column associated with the subject's body was not applied to the head, only that associated with his head and neck. This difference in the exposure may have accounted for the fact that the injury was moderate.

Congestion such as reported in the human experiment just described also has been noted in recent exposure of animal subjects to prolonged acceleration. In two exposures, goat subjects (ref. 37) were reported to have reached the point of strangulation at 8 G's imposed for 15 seconds (fig. 28). Therefore, it appears that the boundary of moderate and severe injury is determined by congestion in the soft extracranial tissues (neck, nose, and eyes) for exposures between 13 G's for 0.6 second and 8 G's for 15.0 seconds.

In addition to the results of reference 37, results of other recent experiments (refs. 37 to 43) indicate that animals exposed to lethal

tailward accelerations die from suffocation instead of intracranial hemorrhages. Experiments cited in reference 29 indicated that animals died from intracranial hemorrhages when subjected to relatively low tailward acceleration. The experiments reported in references 37 to 44, however, contradict the earlier conclusions of reference 29. Consequently, the indicated danger of intracranial hemorrhages from low-G exposures has been attributed by the original investigators in the field (ref. 29) to errors in their original pathological technique (ref. 45). It then appears that the hazard of intracranial hemorrhage caused by tailward impact accelerations is quite small compared with that of other injuries.

Instead of intracranial hemorrhage, the magnitude and duration of rapidly applied tailward accelerations that a properly supported occupant can endure probably is limited by the mechanical strength of the skeleton, as in rapidly applied headward accelerations. When exposures exceed 13 G's and 0.6-second duration, suffocation may occur from congestion in the soft extracranial tissues of the neck.

As shown by figure 28, human tolerance limits possibly may be much greater than the values reported from existing voluntary human exposures. This figure also indicates that the properly supported occupant may survive, with reversible injury, exposures much more severe than the voluntary exposures.

<u>Variation of limits with vehicle onset rate</u>. - The small amount of available onset-rate data is summarized in figure 29. In this figure, the sloping parallel lines indicate onset rates of increasing severity as designated.

The maximum experimental tailward onset rate that human beings have tolerated is 80 G's per second to a level of 8 G's. This exposure was described by the subjects as a hard jolt (ref. 46). The restraining harness in this case was a standard lap strap and shoulder harness.

Subsequent experiments (refs. 32 and 33) have shown that the addition of the lap-strap tie-down strap and reduction of the onset rate to 60 G's per second increased voluntary tolerance to a plateau level of 10 G's. The subjects in these experiments reported no discomfort from the exposure. It was concluded from these exposures that the forces experienced were not only safe for human beings but also were considerably below the injury threshold. These conclusions were later verified by the seven in-flight ejections described in the previous section. The general sensation of the ejection force was reported to be very mild.

Experiments with animal subjects have been made at much greater onset rates. Supine chimpanzees on a high-speed sled (ref. 9) have endured exposure to 22 G's at 1400 G's per second with straps running from their feet over each shoulder and back on the opposite side of their bodies to

tneir feet. These straps were held in place by chest and lap straps (fig. 30). Again, the spine below the shoulders was compressed. This exposure was reported as uninjurious and undebilitating.

Exposure of chimpanzees at the same onset rate (1400 G's/sec) to a peak of 65.5 G's resulted in severe shock, but no other injury was detected. Figure 29 shows a chimpanzee exposure to 48 G's at an onset rate of 1078 G's per second without permanent injury, although severe shock was induced. This subject had sustained a 124-foot-per-second velocity change in 0.24 second. An apparent wide margin of safety thus may exist between the tolerable 80-G-per-second onset rate in human exposures and the 1400-G-per-second onset rate survived in the chimpanzee exposures. It then appears possible that human beings may survive onset rates and plateau magnitudes considerably greater than the present voluntary human tolerance limits.

Variation of limits with amount of body support. - As with accelerations applied in other directions, the arrangement of the restraining harness also affects the severity of tailward acceleration that can be tolerated. Results of the experiments reported in reference 46 showed that the standard seat-belt and shoulder-harness installation did not provide adequate restraint against tailward acceleration of the occupant. The lap strap rotates headward and allows the occupant to leave the seat. Such headward displacement of the lap strap is apparent in figure 31 for a standard lap- and shoulder-strap combination. Because of the lap-strap rotation, the subject was displaced from the seat about 6 inches (ref. 32). Under such circumstances, the lap strap carries only a small part of the load. The shoulder straps carry the remaining part of the load to the shoulders and thence as a compressive load to the spine.

In previous experiments (ref. 31), the headward movement of the lap strap was reduced by moving its attachments forward so that the lap strap was nearly perpendicular to the thighs. With the lap strap so attached, headward rotation of the lap strap around its attachment was prevented. Although this repositioning of the lap strap increased the tolerable level of tailward acceleration to 7.7 G's on the seat, the new position would not be satisfactory in crashes, where forces act upon the seat in many directions.

Results obtained from experiments with the repositioned lap strap just described led to the conclusion that, in ejection seats, the major portion of the tailward ejection force should be applied to the occupant through the lap strap. The major portion of the ejection force is thus transmitted directly to the pelvic structure. When the pelvic structure carries the major part of the accelerating force, the spine is not subjected to the severe compressive stresses that result from the force carried to the shoulders by the shoulder straps.

Further studies (refs. 32 and 33) indicate that the standard seat-belt and shoulder-harness installation could be modified to provide adequate restraint. Addition of a lap-belt tie-down strap as shown in figure 32 would increase the voluntary tolerance to 10 G's on the seat. This strap has been denoted by its function, the lap-belt tie-down strap. Although this strap often is referred to as an inverted "V" strap (refs. 32 and 33), it is not synonymous or interchangeable with the inverted "V" or thigh straps used in the crash harness of reference 12. Comparison of figures 5 and 32 shows the difference between these straps. With addition of the lap-belt tie-down strap, the subject is displaced from his seat only 3 inches (fig. 33, ref. 32), the lap belt is held in place against the pelvic girdle, and the major part of the accelerative force is carried by the pelvis.

DYNAMIC RESPONSE OF OCCUPANT WITH RESPECT TO RAPIDLY APPLIED SEAT ACCELERATION

As previously mentioned, the limits for human tolerance to acceleration presented in this report are based upon a trapezoidal-shaped acceleration pulse measured on the seat pan of a very stiff, strong seat. The effect of onset rate was also based upon this acceleration pulse. This approach to the problem, however, includes the dynamic response of the restraining harness, seat cushions, and so forth, as part of the effect of "onset rate." The dynamic response of the harness and seat, however, should not be included as part of the effect of onset rate, because the resiliency of these components will vary with their design. At present it is not possible to isolate the dynamic response of the harness, because sufficient data are not available. Therefore, it was necessary to present the gross effect of onset rate on the entire elastic chain from the seat pan to the internal organs and define this as the effect of onset rate.

In order to isolate the dynamic response of the particular harness used in any group of experiments, additional measurements should be obtained. The spineward, sternumward, headward, and tailward accelerations of major body parts such as hips, chest, and head should be measured in addition to the longitudinal and vertical accelerations of the seat pan. When these accelerations are known for a series of pulses of progressively increasing onset rate, a dynamic response curve can be drawn for the particular type of restraining harness being used.

A limited amount of such data was available from reference 12 for forces acting in the spineward direction. Using these data, it was possible to plot the major part of a response curve (fig. 34) for a restraining harness made of 3-inch nylon webbing and composed of lap, shoulder, and thigh straps. In this figure the ordinate represents the

maximum acceleration measured on the occupant's chest divided by the acceleration measured on the seat pan. The ordinate thus indicates the amplification or attenuation of the applied acceleration pulse. The abscissa represents the calculated period of an equivalent applied sinusoidal acceleration pulse.

A curve following the general shape of a response curve for a simple elastic system has been faired through the experimental data. In general, the data follow the general trend of the response curve. The curve then shows that with this particular restraining harness the acceleration inflicted on the occupant would be less than the acceleration applied to the seat pan while the period of applied pulse was less than approximately 0.075-second duration. This portion of the curve is based upon few data points; consequently, the value for the duration is approximate. If the applied pulse period were from 0.075 to 0.28 second, the blow inflicted on the occupant would be greater than that applied to the seat pan. At a pulse period of about 0.11 second, a maximum amplification of about 1.6 times the seat-pan magnitude would be applied to the occupant's chest. If the pulse period is greater than 0.28-second duration, the chest and seat acceleration would be essentially equal.

From the values presented in the previous paragraphs, it can be seen that the response curve for this harness arrangement is essentially similar to theoretically derived response curves for simple elastic systems. It thus appears possible to treat this harness-occupant elastic system in the same way as simple elastic systems.

Fragmentary data for impact forces applied sternumward show that amplification can occur also when the person is seated with his back to the accelerating force. The amplification in one run in reference 14 was 2 to 1. In an essentially similar run, the amplification was 1.57 to 1 for an onset rate of 1150 G's per second. In a third run the amplification was 1.25 to 1 for an onset rate of 945 G's per second. When the onset rate was reduced to 550 G's per second, the amplification was negligible. A trend in the dynamic response similar to that for the spineward accelerations is apparent.

For the dynamic-response approach just described to be practical, it must also be possible to establish limits for acceleration tolerance and injury levels, as was done on the basis of the trapezoidal acceleration pulse on the seat pan. This has been done with the data for the spineward dynamic-response curve. The resulting tolerance- and injury-level limits are shown in figure 35. This figure shows the limits for nondebilitating noninjurious exposures and a zone of gradually increasing injury with increasing severity of exposure. In general, this figure is similar to the previously presented tolerance- and injury-level figures except for the generally greater magnitude of the accelerations for the various levels of tolerance or injury. It appears then that this

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approach is practical, but the dynamic-response curves must be determined for each different restraining-harness arrangement and material. These curves can be obtained, however, with accelerations of small magnitude and thus without exposing subjects to injurious magnitudes of acceleration. The tolerance-level curves need be established only once for the various directions of acceleration with respect to the spine, and possibly for the various critical body parts such as hips, chest, and head.

Even though the dynamic response of the restraining harness, cushions, and other padding was obtained, a physiological reaction to onset rate may still remain. The body itself is a link in the elastic chain from the seat to the subject's internal organs. The skeleton, connective cartilage, muscular tissue, and supporting ligaments are all elastic to some degree and transmit accelerative forces to internal organs. These internal organs must certainly be sensitive to accelerative forces. The body itself is thus an elastic system that will have a dynamic response to accelerative forces. Consequently, the onset rate and magnitudes of accelerations externally applied to a person's body will be modified in being transmitted to the internal organs.

Once the human tolerance to the magnitude, duration, and possibly the onset rate of sudden accelerations from various directions and the dynamic response of the harness or supporting cushions have been defined, only the dynamic response of the particular seat design being considered and the accelerations to be expected in various types of impacts are needed for a complete appraisal of the hazard. The dynamic response of the seat can be obtained by the procedure described in reference 47.

VOLUNTARY MUSCULAR RESTRAINT DEMONSTRATED IN LIVE EXPERIMENTS

Occasionally, it is argued that human beings can effectively resist impact forces through voluntary muscular effort. However, experiments show that muscular effort is inadequate for resisting anything but the mildest of impacts.

Reference 48 describes experiments that show human beings unable to move their bodies against forces exceeding 2 or 3 times normal body weight (2 or 3 G's). Effective movement of the body ceased at this level whether the subject was crawling, walking, climbing up a rope or ladder, or rising from the seated position. Strip photographs are shown in reference 48 for subjects crawling, rounding a barrier, and trying to don a parachute under stresses of 1, 2, and 3 G's. Resisting forward rotation around a simple lap strap by voluntary muscular effort does not appear possible, since recorded longitudinal stopping forces in experimental crashes have reached 20 G's in transport crashes and 60 G's in fighter crashes. It further appears that these are not limiting values of possible longitudinal accelerations.

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When a person's body is held by a harness but the arms and legs are unrestrained, he can resist relatively larger accelerative forces by clutching the seat arms or bracing his feet. Studies reported in references 12 and 14 show that, with adequate torso restraint plus vertical or horizontal hand-holds, the arms alone could be supported effectively against spineward and sternumward forces of about 20 G's. Figures 5 and 11 show these types of hand-hold. Note in these figures that there is no arm rest to provide a partial support for the arms. In spineward exposures over 20 G's, hand-holds were abandoned and the arms and wrists were tied as described in references 7 and 12 (fig. 4).

The preceding result is corroborated by two sternumward exposures of about 35 G's reported in reference 12. In both exposures, the subjects were unable to retain a grip on the vertical hand-holds shown in figure 11. Apparently no more experiments were conducted with forces applied sternumward.

4 back

When exposed to headward forces, upward ejection studies have demonstrated that arm position can be maintained against accelerations ranging from 20 to 28 G's with properly designed arm rests. In one sense, muscular restraint is not necessary, because the arm rests carry the inertial forces of the arm; but muscular restraint is needed to hold the arms in position in the arm rests. Arm rests must be designed so that the arm lies in a trough and thus cannot slide off the rest when the headward force is applied. In experimental studies, subjects were instructed to bear down on the rest with forearm and elbows to assure that the arms did not jostle out of the trough. The use of arm rests in relieving the total force on the spinal column has been discussed previously (refs. 11, 21, 26, and 27). An example of an experimental arm rest shaped to hold the forearm in position is shown in figure 22 (reproduced from ref. 26).

Studies of tailward acceleration (ref. 32) indicate that subjects have resisted upward arm loads of 10 to 15 G's. The maximum acceleration in these studies ranged from 10 to 15 G's. Grip on the "D" ring hand-hold was easily maintained during ejection-tower exposures, unless willfully relaxed. Thus, 10 to 15 G's were easily tolerated, and the tolerance limit is probably greater. Upon relaxing the grip on the "D" ring at the seat pan, the subject's hands and arms moved to about the shoulder level.

Mechanical aids for leg and foot restraint have been required for blows from all directions except headward. In tailward seat ejection, the feet could not be held on the foot rest in excess of 4 or 5 G's (refs. 30, 32, 33, and 46). Figure 31 shows feet rising off the foot rest about 4 inches. Simple toe straps provided adequate protection to maximum exposures of about 15 G's. Simple toe straps and canted foot

rests also were found satisfactory for leg restraint against spineward and sternumward stopping forces of 35 G's.

CONCLUSIONS AND RECOMMENDATIONS

Study of the live experiments reported in the literature shows that complete torso and extremity support is the primary variable in increasing survival of rapidly applied accelerations. Therefore, survival of impact forces increases with increased distribution of the arresting force to the entire skeleton, for blows from all directions. The major portion of the accelerating force must be transmitted directly to the pelvic structure and not via the vertebral column. Accordingly, restraints must be designed to support the vertebral column, including the pelvic girdle, as nearly as possible in the normal standing alinement. Unstabilized restraining straps that apply the blow to the soft abdominal walls do not provide maximum body support.

When the conditions of adequate restraint have been met, then the variables, direction, magnitude, and onset rate of rapidly applied acceleration, govern maximum tolerance and injury limits. With respect to direction, tolerance is lower when forces are applied parallel to the spine than when applied perpendicular to the spine. Within the limits of the experiments, the literature shows that the magnitude of tolerable acceleration varies as follows:

(1) Tolerable magnitude decreases as plateau duration increases.

(2) Tolerable magnitude decreases as onset rate increases.

The literature further indicates, from analysis of human survival of falls and from exposure of animal subjects to rapidly applied accelerations, that adequately restrained human beings may survive impact accelerations of greater magnitudes and onset rates than the voluntary tolerance limits.

Recent experiments with animal subjects have produced negligible evidence of the possible occurrence of intracranial hemorrhages from tailward impact accelerations. In other experiments, human subjects have demonstrated that muscular effort is inadequate for resisting anything but the mildest of impacts.

Table IV illustrates conventional human-body restraint and possible increased impact survivability available by use of additional body restraint. The last column of this table shows restraints required for maximum exposures in live experiments.

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Combining all restraints in the third column of table IV shows that adequately stressed aft-faced passenger seats offer maximum complete body support with minimum objectionable harnessing. Such a seat, whether designed for 20-, 30-, or 40-G dynamic loading would include the following:

(1) Lap strap

(2) Chest strap (axillary level)

(3) Full height, integral head rest with winged-back seat to increase headward and lateral G protection

(4) Load-bearing arm rests with recessed hand-holds and provisions to prevent arms from slipping either laterally or beyond the seat back

(5) Leg support to prevent legs' being wedged under the seat (1/2-inch-thick felt padding and special head-rest padding has provided sufficient cushioning for human exposures).

The suggested rigid, aft-faced transport seat could protect occupants against 40-G impacts with moderate or no injury resulting. Moderate injury might be inflicted from headward accelerations exceeding the 20 to 30 G's that now appear tolerable when the occupant is restrained in seated posture as suggested.

For crew members and others whose duties require forward-faced seating, survival of maximum impact forces requires body support as follows:

(1) Lap strap

(2) Shoulder straps

(3) Thigh straps

(4) Lap-belt tie-down strap

(5) Full-height integral head support

With this harness arrangement, shoulder straps always should be slightly looser than lap, thigh, and tie-down strap in order to transmit the major portion of the crash force to the pelvis. Lap straps always should be as tight as comfort will permit, thus eliminating all possible slack. Double-thickness, number 9 drawn nylon straps 3 inches wide have provided sufficient surface area and ultimate strength and elongation characteristics in the maximum human and animal experimental exposures. Extensible fabrics such as undrawn nylon appear extremely hazardous, because